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Effects of hip abductor muscle fatigue on gait control and hip position sense in healthy older adults

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ABSTRACT

We experimentally investigated whether unilateral hip abductor muscle fatigue affected gait control and hip position sense in older adults. Hip abductor muscles were fatigued unilaterally in side-lying position in 17 healthy older adults (mean age 73.2 SD 7.7 years). Hip joint position sense was assessed by an active–active repositioning test while standing and was expressed as absolute and relative errors. Participants walked on a treadmill at their preferred walking speed, while 3D linear accelerations were collected by an inertial sensor at the lower back. Gait parameters, including step and stride time, local divergence exponents and harmonic ratio were quantified. In fatigued gait, stride time variability and step-to-step asymmetry in the frontal plane were significantly increased. Also a significantly slower mediolateral trunk movement in fatigued leg late stance toward the non-fatigued leg was observed. Despite these temporal and symmetry changes, gait stability in terms of the local divergence exponents was not affected by fatigue. Hip position sense was also affected by fatigue, as indicated by an increased relative error of 0.7° (SD 0.08) toward abduction. In conclusion, negative effects of fatigue on gait variability, step-to-step symmetry, mediolateral trunk velocity control and hip position sense indicate the importance of hip abductor muscles for gait control.

1. Introduction

Balance control is essential for successful performance of daily activities and requires adequate sensory integration of visual, vestibular and somatosensory information [1]. Somatosensory information provides feedback about joint position and movement (kinesthesia). Muscle spindles impart the most important proprioceptive input [1]. Rat studies suggested that with aging, the sensitivity of muscle spindles in static and dynamic conditions, the total number of intrafusal muscle fibers and the number of nuclear chain fibers per spindle decrease [2]. Moreover, in older adults, proprioceptive acuity and muscular output may be reduced as a result of an increased fatigability because of loss of muscle strength and central activation failure [3]. Muscle fatigue is defined as an acute impairment in the ability to produce maximum force [4]. In general, muscle fatigue may arise during muscular contractions due to failure at one or more sites along the pathway of force production from the central nervous system to the contractile apparatus [5]. Muscle fatigue has also been associated with decreased proprioception [6,7].

During human gait, hip abductor muscles can stabilize the trunk over the stance leg in mediolateral (ML) direction [8]. In addition, biomechanical modeling showed that the hip abductor muscles apply a moment to accelerate the center of mass (COM) medially toward the swing leg in late stance [9]. Hence, hip abductor weakness might explain impaired gait control in ML direction. For example, following hip replacement surgery, asymmetrical loading has been shown between left and right legs, with a more variable gait pattern [10]. Additionally, the incidence of hip fractures due to sideway falls in older adults might be the consequence of hip abductor weakness and impaired ML balance control [11]. As age-related impairment in hip abduction/adduction torque-time capacity is evident [12], it is important to gain more knowledge on the consequences of hip abductor muscle

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impairment in ML gait control in older adults. One way to experimentally induce muscle impairment is by means of local muscular fatigue, in which the isolated role of hip abductor muscles could be investigated. For instance, hip abductor muscle fatigue is known to increase postural sway velocity in both the sagittal and frontal planes during unipedal stance in young adults, indicating a decline in postural control in static standing [13].

The aim of the present study was to experimentally explore whether unilateral hip abductor muscle fatigue affects gait control in older adults. Since decreased muscle force is the direct effect of fatigue [4], we hypothesized that unilateral hip abductor fatigue changes gait control mainly in ML direction by decreasing stability and symmetry as well as by increasing spatiotemporal variability, possibly because of changes in the trunk COM velocity. The second aim of the study was to investigate the effect of hip abductor fatigue on hip position sense. As muscular fatigue negatively affected proprioceptive acuity in the knee and ankle [7], we hypothesized that it might have the same effect in the hip, possibly contributing to impaired gait.

2. Methods

2.1. Participants

Seventeen healthy volunteers (twelve females; mean age 73.2 SD 7.7 years; height 166.1 SD 9.2 cm; weight 68.2 SD 8.8 kg) with no neurological or musculoskeletal impairments participated in this study. The local ethics committee approved the protocol and participants signed a consent form.

2.2. Assessment of joint position sense (JPS)

First we assessed JPS during hip abduction, for which participants were asked to stand on their unfatigued leg on a 10 cm high block, with their fatigued leg unsupported but aligned with the supported leg (starting position). They were allowed to touch a horizontal bar in front of them at hip height for support. The JPS for hip abduction was assessed unilaterally by an "active–active" and reliable repositioning test [14]. Four trials of hip abduction were performed with target angles randomly varying within a range of 10–40°. Kinematics of the lower limbs were recorded using 6 LED markers (Optotrak, Northern Digital Inc, Waterloo, Canada) at 100 Hz. The markers were attached bilaterally at the apex of the iliac crest, greater trochanter and lateral femur epicondyle to calculate the 3D angle between the vector from the greater trochanter to knee marker and the vector from the greater trochanter to the iliac crest marker. Hip JPS was quantified as the absolute angular error (AE) between target and reproduced angle and the relative angular error (RE) between the two angles with consideration of the direction of the error. A positive relative error indicates an overshoot toward abduction. Because data of the individual trials showed no correlation of AE and RE with the target angle, final values for both AE and RE were obtained by averaging the error over four trials at different target angles.

2.3. Gait and fatigue protocol

The participants walked on a treadmill at preferred walking speed while looking at a wall in front of them and without holding the handrails. A safety harness was used to support full body weight in case of an impending fall. Preferred walking speed was determined by gradually increasing the speed of the treadmill until the participants indicated that the walking velocity was experienced as comfortable. Then, the examiner further increased the speed by 0.5 km/h and checked whether it was experienced as comfortable again. If so, this procedure was repeated until participants indicated that it was too fast. Then, the speed was gradually decreased to the point that participants were sure about their preferred walking speed.

Fig. 1. Set-up to induce hip abductor muscle fatigue, by 30° abduction cycles at 20 repetitions per minute.

After a period of 5 min familiarization and setting the preferred walking speed, they were instructed to walk for 5 min in order to collect 3D linear accelerations of the trunk during walking at a sampling rate of 100 Hz by an inertial sensor (Dynaport Hybrid, McRoberts, The Hague, The Netherlands). The sensor was attached with an elastic belt with Velcro fixation at the lower back at the level of L5.

Then, acute fatigue in the hip abductor muscle group was induced. Participants were placed side lying on a firm mattress with their body and leg straightened (Fig. 1). Randomly, half of the participants were fatigued on their right side and the other half on their left side. A plastic bar was positioned over the participant's foot and set to a height that corresponded to 30° of hip abduction with their knee fully extended. The bar gave the participant a fixed target for the amount of hip abduction required for every repetition and also offered tactile feedback. We instructed participants to raise their leg to touch the plastic bar on each repetition. A metronome was used to pace the lift-and-lower phases, yielding a rate of 20 lifts per minute for all participants. Weight resistance was used to promote the onset of fatigue and was equal to 20% of the participant's total leg moment as estimated from anthropometrical data. Fatigue was defined as the instant at which the subject failed either to reach the target range of motion or went out of sync with the pacing for 3 consecutive repetitions. Immediately after the fatigue protocol, we assessed the subjective exertion level through the Borg CR-10 scale. To ensure maximal effort from each subject, strong verbal encouragement was given throughout the protocol.

Immediately after the fatigue protocol, the JPS post-test was repeated. Since the effect of fatigue can quickly disappear, we repeated the fatigue protocol once more, followed by the post-fatigue walking test.

2.4. Analyses of gait parameters

First, to avoid inaccuracies due to misalignment of the sensor to the anatomical coordinate system, a correctional rotation was made [15]. Then, after low-pass filtering the forward acceleration signal, heel contact (HC) was determined based on the anteroposterior acceleration signal [16]. Mean stride time and variability (standard deviation) was calculated based on HC detection over the first 150 strides.

ML position data were calculated by a double integration of ML acceleration data to discriminate between fatigued and non-fatigued leg HCs, based on an analysis of ML movements of the lower trunk [16]. Fatigued leg stance time was then determined as the time from fatigued leg HC until the following non-fatigued leg HC, whereas non-fatigued leg stance time was determined from

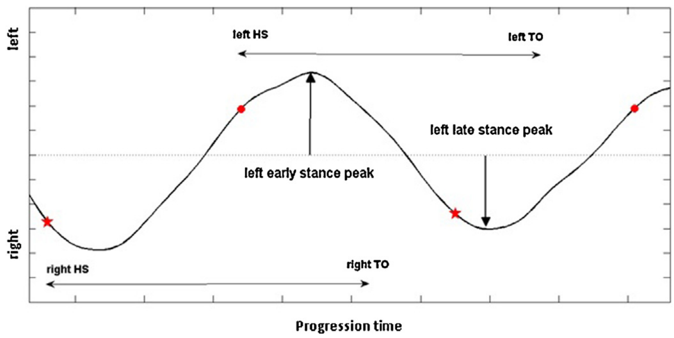

Fig. 2. Trunk velocity in mediolateral direction over time. The stars indicate right heel strike (HS) and the circles indicate left heel HS. Toe-off (TO) happens after the peak which was approximated in the graph. Note that left late stance can be also considered right early stance.

non-fatigued leg HC until fatigued leg HC. Mean stance time was calculated for the fatigued and non-fatigued legs as the average time of 150 fatigued or non-fatigued stance times. Fatigued and non-fatigued stance time variability was defined as SD of the fatigued and non-fatigued stance times.

Harmonic ratio (HR) was quantified as the index of step-to-step symmetry. Stride frequency (number of strides per second) was used to evaluate the harmonic content of the acceleration signals in mediolateral (ML) and anteroposterior (AP) directions. Higher HRs indicate a more symmetric pattern between steps [17].

Based on the first integration of the ML acceleration data, linear ML trunk velocity was calculated. As illustrated in Fig. 2, two velocity peaks could be observed within each stride during the double support phases. The first peak in the velocity profile after ipsilateral HC was defined the early stance peak velocity, while the second peak, just after contralateral HC was considered the late stance peak. During early stance, the peak velocity toward the ipsilateral side mainly reflects the amount of hip abductor control in decelerating the body COM [9], and the peak velocity during late stance reflects the medial acceleration of the body COM by hip abductors. We focused on this latter peak in the late stance phase of both the fatigued and unfatigued leg. Note that an (ipsilateral) late stance peak is actually also the early stance peak of the next (contralateral) step (Fig. 2). Peak velocities of the ML trunk were calculated and averaged over 150 fatigued or non-fatigued late stance periods.

For the calculation of gait stability, local divergence exponents (LDE) were calculated for linear acceleration and angular velocity in ML and AP directions, based on a reconstruction of a 7-dimensional state space with 10 samples (0.1 s) time delay [15]. Rosenstein's algorithm was applied to track the average logarithmic divergences between neighboring trajectories in the reconstructed state space [18]. LDE was quantified as the slope of the first 60 samples (0.6 s) of the divergence curve, which roughly corresponded to one step and it was calculated over equal time-series of 18,000 samples, which roughly corresponded to 150 strides, to ensure that all trials were analyzed similarly.

2.5. Statistical analysis

The normality of the difference in gait parameters and JPS for the non-fatigued and fatigued conditions was checked by visual inspection and Shapiro–Wilks tests. If the differences were normally distributed, we used paired t-tests to determine whether fatigue significantly affected the gait parameters and JPS. A Wilcoxon signed-rank test was used for the outcome variables of which the differences were not from a normal distribution. The t-values from parametric tests and z-values from non-parametric tests were used to determine the Pearson's r as a measure of effect size [19,20]. An r-value higher than 0.1, 0.3 and 0.5 was considered as a small, medium and large effect size, respectively [21]. Statistical analyses were performed with IBM SPSS Statistics 20.0 and the level of significance was set at 0.05.

3. Results

The duration of the first fatigue protocol (before the JPS post-test) was on average 3.00 min (SD 0.82) and the second protocol (before the walking post-test) lasted on average 2.94 min (SD 0.99). At the end of both fatigue protocols, all participants reported a score of 8 or higher on the Borg CR-10 scale.

The average walking speed was 3.63 km/h (SD 0.82). Mean stride time was not significantly affected by fatigue. However, stride time variability was significantly increased when fatigued ($p = 0.02$, $r = 0.56$) (Table 1). Furthermore, the HR in ML direction was significantly different between conditions with an effect size of $r = 0.54$ ($p = 0.01$), indicating less symmetry in the fatigued condition. The peak ML trunk velocity was lower in the fatigued leg late stance phase in the fatigued condition compared with baseline, although not significantly ($p = 0.07$). Finally, none of the gait stability measures were significantly affected by fatigue (Table 1).

No significant difference between the stance times or stance time variability was found between legs in either baseline or

Table 1
Hip position sense and gait parameters (mean with standard deviation SD or median with interquartile range IQR) without or with unilateral hip abductor fatigue. Differences between conditions are indicated by p = values (bold if significant) and effect-sizes (r-values).

	Non-fatigued (baseline)	Fatigued	p-Value	Effect size (r)
Spatio-temporal gait parameters				
Mean stride time (s)	1.09 (SD 0.04)	1.10 (SD 0.04)	0.08	0.42
SD stride time (s)	0.04 (IQR 0.02)	0.05 (IQR 0.02)	**0.02**	0.56
Mean ML peak trunk velocity (m/s) – late stance fatigued leg	0.019 (SD 0.006)	0.017 (SD 0.005)	0.07	0.47
Mean ML peak trunk velocity (m/s) – late stance non-fatigued leg	0.018 (SD 0.005)	0.018 (SD 0.005)	0.78	0.06
Step-to-step symmetry				
HR – ML	2.67 (SD 0.46)	2.45 (SD 0.43)	**0.01**	0.54
HR – AP	3.02 (SD 0.84)	2.97 (SD 0.78)	0.71	0.09
Gait stability				
LDE of ML acceleration	0.81 (SD 0.13)	0.80 (SD 0.10)	0.56	0.15
LDE of AP acceleration	0.81 (SD 0.09)	0.82 (SD 0.11)	0.90	0.03
LDE of AP angular velocity	1.00 (SD 0.12)	1.02 (SD 0.15)	0.56	0.15
LDE of ML angular velocity	1.06 (IQR 0.27)	0.97 (IQR 0.27)	0.55	0.15
Hip position sense				
Absolute error (°)	1.28 (IQR 0.71)	2.04 (IQR 1.33)	**0.02**	0.53
Relative error (°)	0.43 (IQR 1.18)	1.13 (IQR 1.10)	**0.02**	0.52

LDE, local divergence exponents; HR, harmonic ratio; ML, medial lateral direction; AP, anteroposterior direction.

Table 2
Gait parameters (mean with standard deviation SD) between non-fatigued and fatigued legs in baseline and fatigued condition. Differences are indicated by p = values (bold if significant) and effect-sizes (r-values).

	Non-fatigued leg	Fatigued leg	p-Value	Effect size (r)
Baseline				
Mean stance time (s)	0.55 (SD 0.03)	0.55 (SD 0.02)	0.98	0.00
SD stance time (s)	0.043 (SD 0.02)	0.045 (SD 0.04)	0.66	0.1
Mean ML peak trunk velocity (m/s) – late stance	0.018 (SD 0.005)	0.019 (SD 0.006)	0.66	0.11
Fatigue condition				
Mean stance time (s)	0.55 (SD 0.02)	0.56 (SD 0.03)	0.13	0.37
SD stance time (s)	0.045 (SD 0.04)	0.046 (SD 0.02)	0.53	0.2
Mean ML peak trunk velocity (m/s) – late stance	0.018 (SD 0.005)	0.017 (SD 0.005)	**0.007**	0.61

fatigued conditions (Table 2). However, in the fatigued condition, a significant difference between legs for the late stance peak ML trunk velocity was observed, indicating a significantly lower trunk velocity peak in the fatigued leg late stance (p = 0.007).

With respect to the JPS test, the mean target angle during the four trials of both conditions was 23° (SD 7°). Fig. 3 shows the amount of absolute error over hip abduction angles for all trials of each participant in the fatigued condition. When hip abductor muscles were fatigued, the absolute errors in reproducing the target angles in the JPS test were 0.76° (60%) larger than for the non-fatigued condition (p = 0.02, r = 0.53) (Table 1). The positive relative errors indicated that participants mainly overshot the target angle in the fatigued condition toward abduction by on average 0.70° (p = 0.02, r = 0.52) (Table 1).

4. Discussion

We investigated whether unilateral hip abductor muscle fatigue affects gait control and hip joint position sense in older adults. We found significant effects of fatigue on the gait pattern, as participants showed more stride time variability, less step-to-step symmetry in ML direction, and lower ML trunk velocity in fatigued late stance. Moreover, hip joint position sense acuity was significantly decreased when hip abductor muscles were fatigued.

4.1. Effect of hip abductor fatigue on gait

Stride time variability was significantly increased by fatigue, while mean stride time was not affected. More importantly, ML step-to-step symmetry during walking was lower in the fatigued condition. This ML asymmetry might, at least partly, be explained by the significantly slower ML trunk movement in fatigued late stance toward the non-fatigue leg. Hip abductor muscles serve to decelerate lateral trunk velocity toward the ipsilateral side just after heel strike and accelerate the trunk toward the contralateral side during "push-off", as described by Pandy and colleagues [9]. The velocity generated during push-off was consistently decreased with unilateral fatigue, which could explain the decreased gait symmetry. The comparison within trials between the fatigued and non-fatigued leg confirmed that the asymmetric pattern occurred mostly due to the fatigued leg, as peak trunk velocity in fatigued late stance was systematically lower in all individuals, except one subject who appeared to have an irregular trunk COM pattern during walking. Possibly decreased hip abductor muscle force because of fatigue negatively affected the medially directed component of the push-off by the fatigued leg toward the non-fatigued side in late stance, which would result in a slower trunk movement in late stance. Additional trunk kinematic and kinetic analyses can help to further explain exactly where and how the deviations from symmetry occur [17].

Despite changes in gait parameters, fatigue did not affect gait stability in terms of the local divergence exponents, suggesting that the body as a system was able to deal with the small internal perturbations (i.e. the increased sensory-motor noise) due to fatigue. Toebes and co-workers also did not find significant effects of unilateral knee extensor muscle fatigue on gait stability although the initial mechanical resistance against perturbations in the ML direction was reduced [22]. Apparently, all of our fit and healthy participants [23] were able to use compensatory strategies to control their gait stability, irrespective of unilateral hip abductor muscle fatigue.

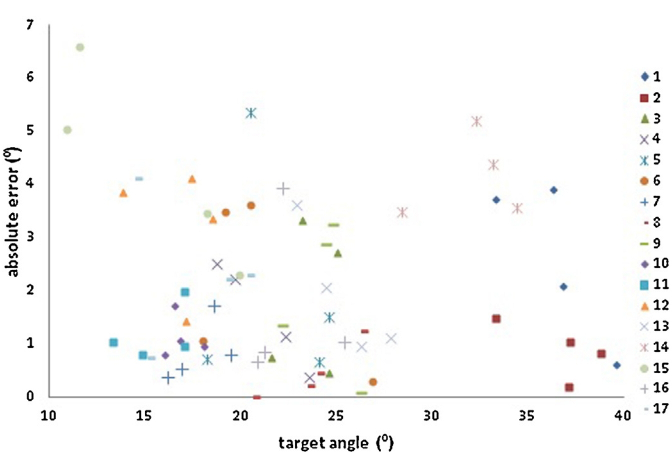

Fig. 3. The amount of absolute error over hip abduction angles for all trials in each participant (1–17) during the JPS test in the fatigued condition.

4.2. Effect of hip abductor fatigue on hip position sense

The decrease in hip joint repositioning due to fatigue was reflected in an increased absolute error, suggesting dysfunction of muscle spindle afferents due to fatigue. In addition, we observed a systematically more positive relative error toward abduction. It is, however, difficult to explain the mechanism behind the systematic abduction error. Fatigue can increase the threshold of muscle spindle discharge and change alpha–gamma co-activation, and consequently the output of the peripheral proprioceptive system [24], but these phenomena do not clearly point to a systematic error toward abduction or adduction. Our results are, nevertheless, in line with previous studies that found overestimation of knee angle due to fatigue [6] as well as higher absolute errors in the knee [25] and ankle [7], using different JPS tests.

Although we did observe a declined JPS after the hip abductors were fatigued, the experimental protocol did not enable us to explain what role, if any, this reduction in hip position sense had on gait parameters, as fatigue negatively affects both muscle force and proprioception. Further investigations are required to examine the relative contribution of hip abductor proprioception and muscle strength in gait control.

4.3. Limitations

We standardized the fatigue protocol by a fixed pace and hip abduction angle with weighted resistance, but each participant nevertheless reached their fatigue level at different times, possibly at different fatigue levels. Also, these fatigue effect might not have sustained long enough throughout the walking session to result in significant effects on all gait parameters. Unfortunately, we did not measure muscle activity patterns or muscle force before and after the fatigue protocol to quantify the level of fatigue over the fatigue protocol or throughout the walking session, which could indicate the degree of recovery from fatigue. We furthermore avoided extreme small or large target angles in our JPS protocol, as previous studies [26,27] suggested that the JPS test might be unreliable in extreme angles, either at the beginning or the end of the range of movement. Therefore, the range of the target angles in the JPS is larger than the normal hip abduction angle during gait. Yet, larger angles in our defined target angles did not lead to greater errors (Fig. 3), suggesting that the JPS results reliably discriminated the effect of hip abductor fatigue on proprioceptive acuity of this muscle group, irrespective of the target angle. Therefore, it might be plausible to generalize the JPS findings for other conditions such as gait.

5. Conclusion

Following a protocol to induce unilateral hip abductor fatigue, healthy older adults exhibited increased stride variability, decreased ML symmetry, and a lower peak ML trunk velocity in the fatigued leg late stance phase during treadmill walking, and a reduced proprioceptive acuity in hip abduction position repositioning.

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Conflict of interest statement

The authors declare that there are no conflicts of interest.

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